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Evaluation of the different sleep-disordered breathing patterns of the compressed tracheal sound

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HIGHLIGHTS

- Even if sustained partial obstruction is found to be common among sleep-disordered breathing (SDB) patients, it is not easily observed in conventional polysomnography recordings.
- Oesophageal pressure decreased and the tracheal sound presented most high frequency components in the 1001–2000 Hz frequency band during the sustained partial obstruction.
- The presented compressed tracheal sound analysis seems to provide an easy and reliable method for screening not only apneas or hypopneas but also sustained partial obstruction.

ABSTRACT

Objective: Suitability of the compressed tracheal sound signal for screening different sleep-disordered breathing patterns was evaluated. The previous results suggest that the plain pattern in the compressed sound signal represents mostly normal, unobstructed breathing, the thick pattern consists of periodic apneas/hypopneas and during the thin pattern, flow limitation in the nasal cannula signal is abundant.
Methods: Twenty-seven patients underwent a polysomnography with a tracheal sound and oesophageal pressure monitoring. The tracheal sound data was compressed and scored visually into three different breathing patterns. The percentage of oesophageal pressure values under -8 cm H_2O, the minimum pressure value and the average duration of the breathing cycles were extracted from 10-min episodes of those plain, thick and thin patterns. In addition, the spectral contents of the tracheal sound during the different breathing patterns were evaluated.
Results: The percentage of time when the oesophageal pressure negativity increased was highest during the thin pattern and lowest during the plain pattern. In addition, the thin pattern presented most high frequency components in the 1001–2000 Hz frequency band of the tracheal sound.
Conclusions: The results confirmed our previous findings that both the thick and thin patterns seem to consist of obstructed breathing, whereas during the plain pattern the breathing is normal, unobstructed.
Significance: Most screening methods for sleep-disordered breathing reveal only periodic apneas/hypopneas, but with the compressed sound signal the sustained partial obstruction can be estimated as well.

1. Introduction

Obstructive sleep apnea (OSA) is common; its prevalence is estimated to be 2–4% (Young et al., 1993). However, in addition to apneas and hypopneas revealing OSA, sleep recordings consist of other sleep-disordered breathing patterns, as well. These patterns include short flow limitation events associated with an increase in negative oesophageal pressure as well as an arousal, as in the upper airway resistance syndrome (UARS), or sustained low limitation periods, lasting from several minutes up to tens of minutes (Hernandez et al., 2001; Tenhunen et al., 2009). The sustained partial obstruction detected by a sleep mattress sensor has been found to be common especially among postmenopausal

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women (Polo-Kantola et al., 2003). According to our recent work, sustained partial obstruction seems to be associated with somnolence and presumably also with depressed mood (Tenhunen et al., 2013).

The need for sleep studies is vast, and various systems either with visual, automated or semi-automated methods has been developed to simplify and speed up the recording procedure. One of the methods used to study respiratory events during sleep is the analysis of tracheal sound. It has been developed to detect sleep apnea and other respiratory events among both in adults and children (Krumpe and Cummiskey, 1980; Beckerman et al., 1982; Cummiskey et al., 1982; Sanna et al., 1991; Nakano et al., 2004; Kulkas et al., 2009). Also developmental works for a diagnostic assessment and portable screening devices have been done (Lugaresi et al., 1983; East and East, 1985; Hida et al., 1988). However, collecting only the events resembling apneas/hypopneas ignores periods with sustained partial obstruction. We have previously presented a visual analysis method to evaluate nocturnal tracheal sound (Rauhala et al., 2008). The method utilises heavy data compression. The resulting visual representation of the compressed tracheal sound signal reveals three distinct sound patterns. The plain sound pattern seems to represent mostly normal breathing with low sound amplitude levels, the thick sound pattern consists of intermittent alternation of sound amplitude levels associated with apneas and hypopneas, and the thin sound pattern, in which the sound amplitude level is continuously high and flow limitation in the nasal pressure transducer is abundant.

In the present work, we continue the evaluation of compressed tracheal sound patterns. We study the changes in the oesophageal pressure values associated with different compressed sound patterns. In addition, we evaluate the spectral content of the raw tracheal sound signal during these compressed patterns. Our hypothesis is that during the plain sound pattern the oesophageal pressure (pESO) values remain normal and that during the thick and the thin patterns, the pESO negativity increases indicating upper airway obstruction. As along with increasing obstruction, the spectrum of the tracheal sound signal consists more of higher frequencies (Rao et al., 1990; Yonemaru et al., 1993; Pasterkamp et al., 1996; Kaniusas et al., 2005; Herzog et al., 2008; Michael et al., 2008), we hope to find out that during the plain sound pattern, the spectrum of the tracheal sound consists mostly of lower frequencies. During the thick and the thin patterns higher frequencies would be more common.

2. Methods

2.1. Subjects and recordings

Twenty-seven consecutive patients (22 male, 5 female) referred to the Sleep Unit of Pirkanmaa Hospital District volunteered to participate in this study. The patients were studied due to the suspicion of sleep-disordered breathing. All of the subjects gave their written consent. The study was approved by the Ethical Committee of Pirkanmaa Hospital District.

The sleep recording montage consisted of six EEG derivations (F3-A2, F4-A1, C3-A2, C4-A1, O1-A2, O2-A1), two EOG channels, submental and anterior tibialis muscle EMG, body position, pulse oximetry, Emfit sleep mattress and electrocardiogram. Thoracic and abdominal respiratory movements were recorded with inductive belts. Airflow was measured by nasal pressure transducer and thermistor. Recordings were performed with the Embla N7000 (Embla®, Natus Medical Inc., USA) and the Somnologica Studio 3 software (Medcare®, Iceland) simultaneously with the oesophageal pressure (pESO) measurement (Reggie, Camtech AS, Norway) and tracheal sound recording. These external device data were converted into Embla data format and their temporal synchronization was confirmed before the analysis (Kulkas et al., 2008; Tenhunen et al., 2009). A sampling rate of 11,025 Hz was used for tracheal sound, 2 Hz for pulse oximetry, 10 Hz for respiratory movements and 20 Hz for oesophageal pressure, 500 Hz for ECG, and 200 Hz for all other signals. The pESO measurement is invasive but the signal has some advantages, for example, it shows breathing efforts both during normal and obstructed breathing (including obstructive apneas) (Reda et al., 2001).

The tracheal sound recordings were performed with an electret microphone (Panasonic WM-60A, Matsushita Electric Industrial Co, Ltd, Kadoma Osaka, Japan). The microphone's conical air cavity is 25 mm in diameter and 3 mm in depth. The sensitivity of the microphone is 10 mV/Pa and the frequency range in the free field is 20 Hz–20 kHz, ±2 dB (Sovijarvi et al., 1998). The microphone was attached to the skin in the suprasternal notch as in our previous studies (Rauhala et al., 2008; Tenhunen et al., 2009; Kulkas et al., 2010). The tracheal sound signal was amplified and high-pass filtered with a cut-off frequency of 50 Hz and fed into a Sound Blaster Audigy 2 NX sound card (Creative Labs, Singapore) for a 24 bit A/D conversion followed by an USI-01 isolator (MESO, Mittweida, Germany) providing galvanic isolation between the patient and the recording equipment. The obtained tracheal sound signal provides the raw data from the sounds recorded over the trachea.

2.2. Visual scoring

Nocturnal polysomnography recordings were scored into the sleep stages according to the standard criteria, and the apnea-hypopnea index (AHI) was calculated according to the hypopnea rule 4B (Iber et al., 2007). Microarousals were scored according to the criteria of the American Sleep Disorders Association (American Sleep Disorders Association (ASDA), 1992).

For the visual analysis of the different sound signal patterns the tracheal sound signal data were reduced as in our previous work (Rauhala et al., 2008). In the reduction procedure only the maximum and minimum sound signal values of each consecutive 15-s epoch was taken. This compressed sound information (four maximum and minimum samples per minute) was used for the visual analysis. The compressed tracheal sound signal was scored into four different categories: a plain signal curve close to zero, a thin signal curve deviating clearly from zero and a thick, highly varying signal curve. The fourth category consisted of the periods of signal periods of low quality (not-scored periods). After that, one 10-min representative episode of plain, thin and thick tracheal sound pattern was selected for present analysis from each subject, whenever present. This selection was performed by two researchers working together to assure the validity of the selected episodes. An example of each type of compressed sound pattern is shown in Fig. 1a, and examples of the raw signals (tracheal sound, pESO, and nasal flow) for 3-min, part of each 10-min pattern, are presented in Fig. 1b–d.

2.3. pESO analysis

A special software was developed to import the pESO signal to the Somnologica software. The oesophageal pressure value of -8 cm H_2O is considered the threshold between obstructive and non-obstructive events when using the Reggie device (Oeverland et al., 2005). This threshold was used to differentiate increased respiratory effort and normal breathing. As the first quantity, the percentage of time with pESO values of -8 cm H_2O or less was calculated during each 10-min compressed sound pattern. This quantity also was used in our previous work validating Emfit sleep mattress performance (Tenhunen et al., 2011). In addition, the most negative oesophageal pressure value (pESOmin) for each 10-min sound pattern was collected automatically.

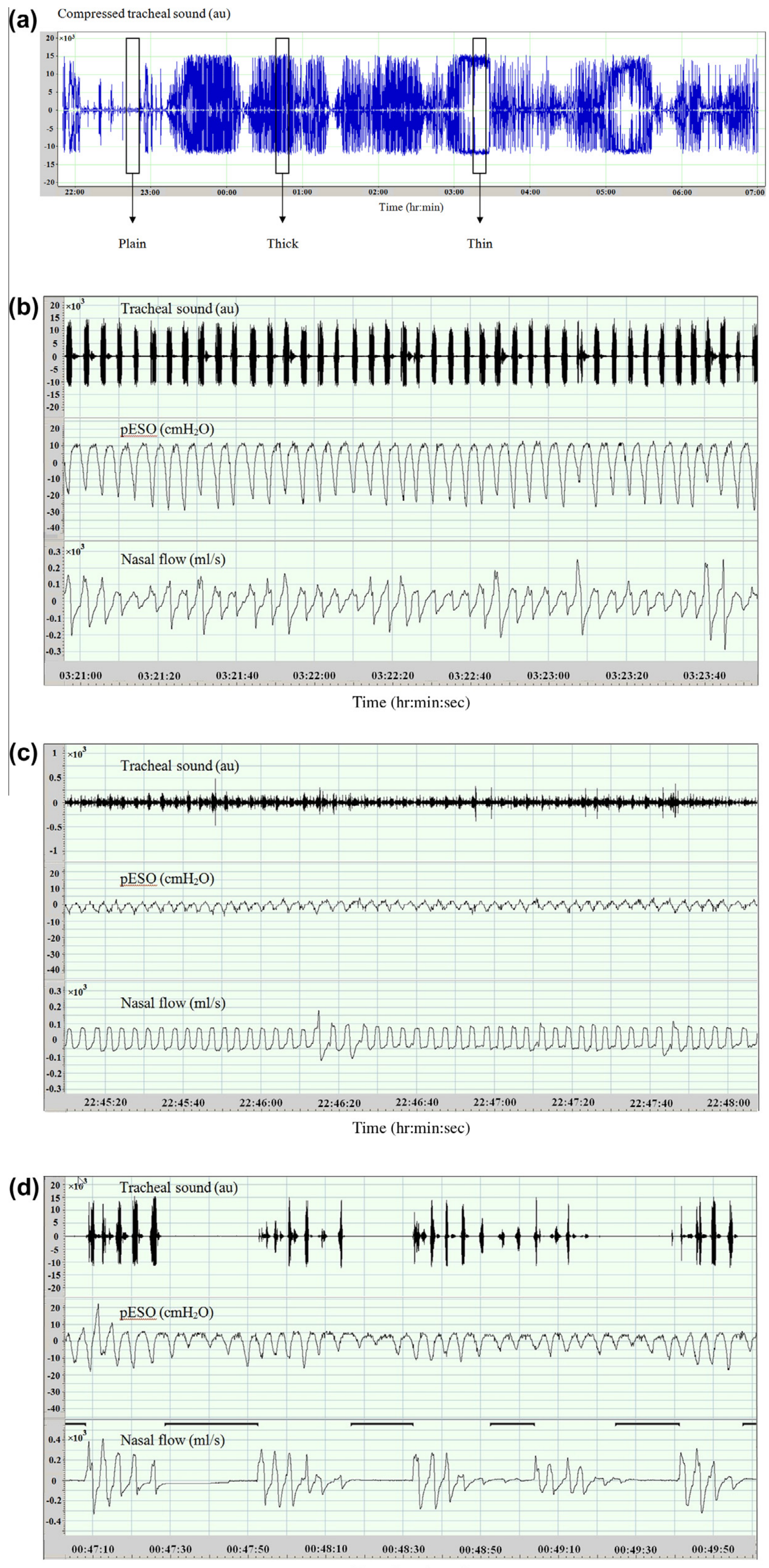

Fig. 1. (a) An example of the compressed tracheal sound signal of one sleep apnea patient (AHI 75/h). The visually selected 10-min long plain, thick, and thin patterns are indicated with narrow rectangles. The plain sound pattern presents low sound amplitude levels (indicating normal breathing), the thick sound pattern shows varying sound amplitude levels (associated with apneas and hypopneas), and during the thin sound pattern, the sound amplitude level is continuously high (suggesting sustained partial respiratory obstruction). The arbitrary unit of amplitude is denoted as 'au'. (b–d) 3-min periods of the tracheal sound, oesophageal pressure (pESO), and linearized nasal flow inside each compressed breathing patterns. (b) Thin breathing pattern; the tracheal sound signal shows repetitive loud sounds throughout the 3-min section. The pESO signal shows increased breathing effort with large decreases in value far below −15 cm H_2O, and the nasal pressure trace shows flow limitation. (c) Plain breathing pattern; Quiet breathing sounds with low sound amplitudes, and normal pESO values with round flow shapes are seen. (d) Thick breathing pattern; intermittent high amplitude tracheal sounds with negative pESO swings during apneas/hypopneas (marked with lines). The arbitrary unit of sound amplitude is denoted as 'au'.

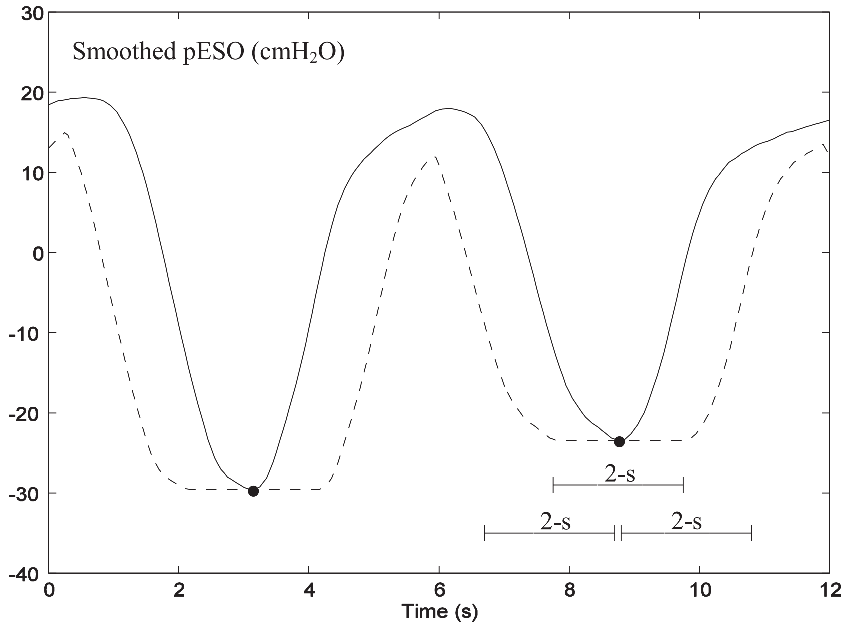

Fig. 2. An example of the smoothed pESO signal and the automated identification of the local minimum points. The smoothed pESO signal is drawn with a solid line and the 2-s minimum curve with a dashed line. The 2-s long segments used in the processing are also illustrated. Identification of one local minimum point (marked with a dot) can be seen at the time-instance of 8.75 s, and a previous one at the time-instance of 3.15 s.

In the further quantification of the pESO signal, the average duration of the breathing cycles inside each 10-min sound pattern was computed. At first, a smoothed version of the pESO signal was extracted by using sliding 1-s mean filtering. Then, going through the smoothed pESO signal, a local minimum point was identified at each time-instance where the three following conditions were met. Firstly, the current sample equaled the minimum value of the samples in the 2-s long segment centered at it (Fig. 2). Further, the current sample was lower than the average values of the samples in the preceding 2-s and following 2-s long segments. The local minima were visually verified. Finally, the average duration of the breathing cycles was obtained by dividing the 10-min epoch by the number of the identified local minima (representing the total number of breathing cycles).

2.4. Spectral analysis of tracheal sound

One-second segmenting was used in the spectral analysis of the raw tracheal sound signal. After mean removal, each segment was weighted with a Hanning window function of respective length. The discrete Fourier transform provided a complex-valued spectrum, which was scaled to the corresponding amplitude spectrum as presented previously (Huupponen et al., 2008). The frequency resolution of the spectrum was 1 Hz as no zero padding was used. The Welch spectrum of the 10-min episode was formed with 50% overlapping. This Welch spectrum enabled examination of the overall spectral content of the raw tracheal sound signal (amplitude values along frequency axis) during the 10-min episode. From the Welch spectrum, mean amplitude in each of the five frequency bands was then determined (50–1000 Hz, 1001–2000 Hz, 2001–3000 Hz, 3001–4000 Hz, and 4001–5512 Hz) for each 10-min episode. In addition, the percentage proportions of the sound amplitudes in the above-mentioned bands as shares of 50–5512 Hz were extracted for each 10-min episode.

2.5. Statistics

Statistical analyses were performed with the SPSS for Windows version 21.0® software (SPSS Inc.). Non-parametric tests were used, as all the variables were not normally distributed. The Friedman test was used to compare the multiple dependent variables and the Wilcoxon signed-rank test was utilized in the post hoc analyses. In the post hoc analyses, appropriate Bonferroni corrections were used. *P*-values <0.05 were considered statistically significant.

3. Results

The sleep parameters derived from the sleep recordings of the 27 subjects are shown in Table 1. Six patients had no marked findings in their polysomnography, one patient had an excessive amount of periodic leg movements and the rest of the patients had OSA (7 mild OSA, 3 moderate OSA, and 10 strong OSA). Sixteen subjects out of 27 had a 10-min plain episode. Nineteen subjects

Table 1
Sleep parameters of the 27 patients.

	Median	Min	Max
TST[a] (min)	404.0	233.5	542.5
SEI[b] (%)	81.7	65.3	95.1
SL[c] (min)	15.5	1.0	67.0
REMlat[d] (min)	171.0	44.0	372.0
% N1[e]	9.8	2.5	35.1
% N2[e]	64.6	41.3	80.5
% N3[e]	11.1	0.8	39.9
% REM[e]	14.6	0.0	24.1
ARI[f]	22.8	6.5	59.2
AHI[g]	13.9	1.5	88.0
ODI4[h]	6.0	0.7	63.0
%SaO_2min[i]	86.5	77.0	94.0
% SaO_2mean[j]	94.0	90.9	96.5

[a] Total sleep time.
[b] Sleep efficiency index.
[c] Sleep latency.
[d] Latency to REM (rapid eye movement) sleep.
[e] Percentage of sleep stage (N1-REM) referred to TST.
[f] Arousal index, number of cortical arousals per hour of TST.
[g] Apnea-hypopnea index, apneas and hypopneas per hour of TST.
[h] Oxygen desaturation index, number of desaturations ⩾4% per hour of TST.
[i] Minimum oxygen desaturation.
[j] Mean oxygen desaturation.

had a thick sound episode and another 19 subjects had at least one 10-min thin episode.

3.1. Quantities of pESO

The resulting percentages of time with increased respiratory efforts (pESO $\leqslant -8$ cm H_2O) during the different compressed sound patterns are depicted in Fig. 3. This percentage was highest during the thin pattern (median 21.4%, range 3.5–31.9%), and the difference was statistically significant as compared to the plain sound pattern (0.03%, 0–1.0%, $p = 0.01$) and to the thick pattern (9.6%, 3.9–20.1%, $p = 0.002$). During the plain sound pattern, the percentage was lowest, also showing significant difference from the thick pattern ($p = 0.02$).

The obtained most negative pressure values (pESOmin) during the 10-min sound patterns are presented in Fig. 4. The most normal (near -8 cm H_2O) pESOmin-value was found during the plain pattern (median -9.0 cm H_2O, range -5.9 to -19.9 cm H_2O), and showed statistically significant difference from both the thin sound pattern (-23.9 cm H_2O, -11.3–-42.7 cm H_2O, $p = 0.02$) and from the thick pattern (-22.4 cm H_2O, -13.0–-52.2 cm H_2O, $p = 0.045$).

The results of the third pESO quantity analysis; the average duration of the breathing cycles during the different compressed sound patterns, are shown in Fig. 5. The average duration of the breathing cycles was longest during the thin pattern (median 4.1 s, range 3.2–5.8 s). It differed significantly from the plain sound pattern (3.2 s, 1.9–4.0 s, $p = 0.01$). During the thick pattern (3.8 s, 2.6–6.4 s), the average duration of the breathing cycles was between the other two patterns, but it did not differ statistically significantly from either.

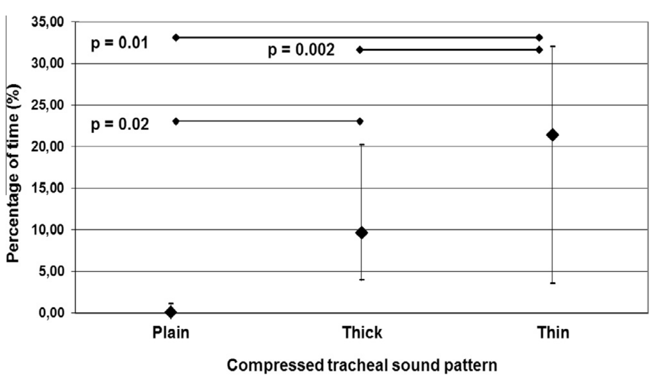

Fig. 3. The percentage of time with increased respiratory efforts (pESO $\leqslant -8$ cm H_2O) during the different compressed sound patterns. Median percentage values with respective ranges are presented. The percentage of time with increased respiratory efforts was highest during the thin pattern and lowest during the plain pattern. The statistically significant comparisons are marked with horizontal lines.

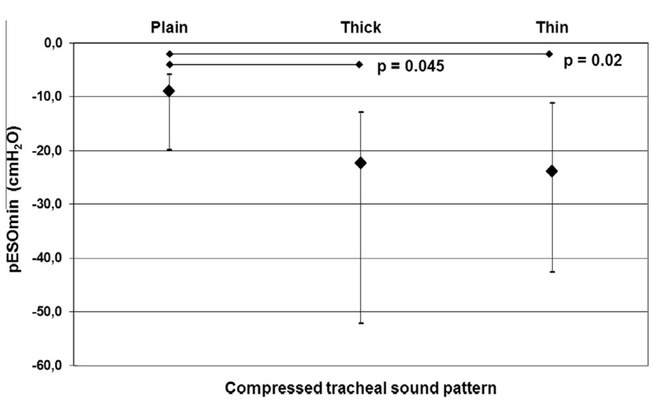

Fig. 4. The most negative oesophageal pressure values (pESOmin) during the 10-min sound patterns. Median values with ranges are presented. The most normal (near -8 cm H_2O) values are observed during the plain sound pattern. During the thin and the thick sound pattern, the values are indicative of disturbed breathing. The statistically significant comparisons are marked with horizontal lines.

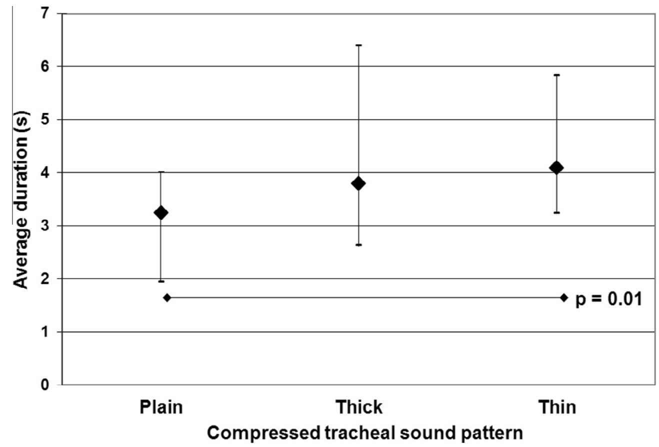

Fig. 5. The average duration of the breathing cycles during the different compressed sound patterns. Median values with ranges are presented. The thin pattern presented the highest duration, differing the most from the plain pattern, this statistically significant comparison is marked with a horizontal line.

3.2. Spectral content of tracheal sound

The obtained tracheal sound amplitudes in the five frequency bands are presented in Fig. 6. The amplitude was the highest during the thin pattern and lowest in the plain pattern showing statistically significant differences in every frequency band (all p-values were <0.05).

The resulting percentage proportions of the spectral amplitudes in the five bands (median values) are depicted in Fig. 7. The thin pattern produced the lowest percentage proportion value in the frequency band of 50–1000 Hz (median 81.1%, range 51.6–97.1%), differing statistically from the thick pattern (90.6%, 53.0–96.5%, $p = 0.02$) but not from the plain (93.8%, 42.8–96.9%). In the frequency band of 1001–2000 Hz, the thin pattern provided the highest percentage proportion value (9.7%, 1.2–17.7%) and differed

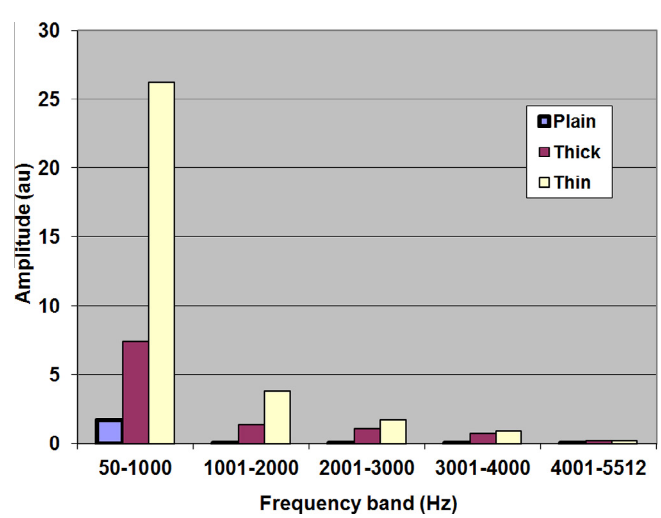

Fig. 6. The tracheal sound amplitudes in the five frequency bands during the three different compressed sound patterns. Median values are drawn with bars. In all frequency bands, the amplitude was highest during the thin pattern and lowest during the plain pattern (all p-values <0.05).

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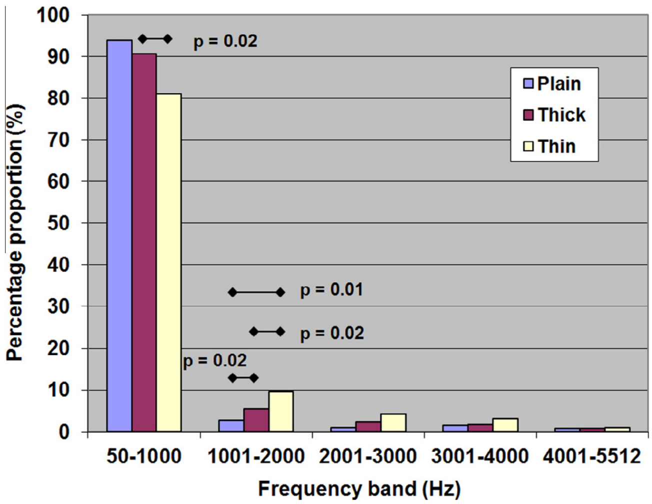

Fig. 7. The percentage proportions of the tracheal sound amplitudes in the five frequency bands during the three compressed sound patterns. Median values are drawn with bars. In the first frequency band (50–1000 Hz), the percentage proportion was lower during the thin pattern than the thick pattern. The percentage proportion in the second frequency band (1001–2000 Hz) was highest during the thin pattern and the lowest during the plain pattern. These findings indicate increased high-frequency spectral content during sustained obstruction. The other three frequency bands showed no statistically significant differences.

significantly from the plain pattern (2.7%, 1.5–15.9%, $p = 0.01$) and the thick pattern (5.4%, 2.2–16.2%, $p = 0.02$). The thick pattern also differed significantly from the plain pattern ($p = 0.02$). The other three frequency bands showed no statistically significant differences.

4. Discussion

The analysis of the tracheal sound might provide an easy and comfortable tool for the evaluation of nocturnal breathing. So far the studies of tracheal/snoring sounds have mainly concentrated on technical solutions and detecting apneas and hypopneas (Cummiskey et al., 1982; Van Brunt et al., 1997; Nakano et al., 2004; Rauhala et al., 2008; Kulkas et al., 2008). Less attention has been paid to the study of sustained partial obstruction. However, among our sleep laboratory patients, the prevalence of patients with sustained partial obstruction was found to be 10% (Tenhunen et al., 2013) and it has been found to be even more common among elderly female patients (Polo-Kantola et al., 2003; Anttalainen et al., 2007).

The results of our previous study on tracheal sound suggested that the presented visual compressed tracheal sound analysis might provide a reliable and efficient tool for screening both periodic apnea/hypopnea breathing and sustained partial obstruction (Rauhala et al., 2008). We introduced the three distinct patterns of tracheal sound signal: plain, thick and thin patterns. The patterns were easily recognized visually. During the plain pattern the sound amplitude levels remained low and the flow shapes detected by the nasal flow pressure transducer were mostly round shaped, thus indicating normal breathing. However, in that study some subjects had apneas or hypopneas during the plain sound periods. Therefore, the assumption that plain sound patterns reflect normal breathing needed further examination. In the present work, the oesophageal pressure values turned out to be in the normal range during the plain breathing pattern. This strengthens our hypothesis that the plain pattern represents normal, unobstructed breathing. It might be that the respiratory events found in the previous study represent normal stage dependent respiratory fluctuation without increased effort.

In our previous study, we discovered that the thick pattern represents highly variable sound levels (Rauhala et al., 2008). It seemed that the share of the thick sound patterns could be used to disclose OSA, as the share of the thick sound patterns correlated well with AHI (Rauhala et al., 2008). The thin sound pattern consisted of abundant flow limitation in the nasal cannula signal. This is why we assumed that both the thick and the thin patterns reflect obstructed breathing. The results of the present study support our assumption, oesophageal pressure values were markedly decreased during both the thick and the thin patterns. Another support for the obstructive nature of the thin pattern stems from our finding that breathing cycle durations lengthened during the thin pattern, as it has been presented that the increase in the duration of breathing cycles is related to the increase in respiratory obstruction (Stoohs and Guilleminault, 1991; Brack et al., 1998).

The other hypothesis of the present work was that if the thin and the thick patterns consist of obstructed breathing, tracheal sound signal during them should be composed of higher frequencies than normal breathing. The assumption that the higher tracheal sound frequencies represent obstruction is supported by earlier studies (Rao et al., 1990; Yonemaru et al., 1993; Pasterkamp et al., 1996; Kaniusas et al., 2005; Herzog et al., 2008; Michael et al., 2008). Here we present that the thin pattern contained tracheal sound components of a higher amplitude than the plain and the thick pattern. During periodic apnea-hypopnea episodes (thick pattern) the breathing is frequently interrupted and there are almost silent periods between the apnea events. On the contrary, during the thin pattern (sustained flow limitation), repetitive loud sounds lasting prolonged periods are observed (Rauhala et al., 2008). In snoring the higher acoustic intensity has been found to reflect more severe respiratory obstruction (Lugaresi et al., 1983; Itasaka et al., 1999; Wilson et al., 1999). This is in line with our finding that sound amplitude was high during both thick and thin patterns. However, according to our earlier (Tenhunen et al., 2009) and the present study the mere sound intensity does not explain the differences of percentage proportions of the sound amplitudes in the different frequency bands. The tracheal sound during the thin pattern was composed proportionally more of high frequency components in the 1001–2000 Hz band than the plain or thick patterns. The proportional amplitude in that band was lowest during the plain pattern. As higher frequency components in the breathing sounds reflect obstruction, our findings suggest that both the thick and the thin patterns represent inspiratory obstruction. It is supposed that during the thin pattern, upper airway constriction is not as total as in periodic breathing (thick pattern) which might enable the airflow to interact more with the vocal tract and the mouth cavity, where more high frequency components of sound are found to be produced (Narayanan and Alwan, 2000). This might be the origin of the high frequency components.

Different computer-based sleep analysis methods have being developed to help the analysis of large amounts of sleep data. In clinical neurophysiology compression of the raw data is often used to facilitate and speed up visual review procedure (Agarwal et al., 1998; Hellstrom-Westas et al., 2006). We think that an important goal is to develop new measures to provide additive information, which is not routinely evaluated. Such methods might provide complementary information to the conventional analysis of sleep recordings and enhance the diagnostic accuracy in detection of the clinically significant sleep disorders. The compressed tracheal sound signal visualizes easily the nocturnal respiratory activities in one overview.

In conclusion, we suggest that the visual compressed tracheal sound analysis would provide a promising method for screening both obstructive apneas/hypopneas and sustained partial obstruction. Many other screening devices can be used to distinguish apneas from normal breathing, but with the presented compressed

analysis, a third pattern with sustained obstruction can also be observed. The device, with a sensitive, but small microphone is easy-to-fix, which makes it suitable for ambulatory sleep recordings.

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